

Non Invasive Measurement and Analysis of Cardiac Output for Different Age Group using PPG Sensor

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ABSTRACT

This study provides analysis of reflectance type PPG (Photoplethysmography) sensor using mean square spectrum. Power spectrum shows that absorption of deoxyhaemoglobin and oxyhaemoglobin is maximum for 660nm and 860nm respectively. The reflectance type PPG sensor was designed for two sources of wavelength are 660nm and 860nm and OPT 101 as a detector. Data collection was done for 10 specimen including female and male in the age group of 20 to 45 years. MSS (Mean square spectrum) reflects the power in the signal at a given frequency. The MSS of a signal is the Fourier transform of that signal's autocorrelation. The acquisition of PPG signal was done using microcontroller with 10 bit ADC and was processed in Matlab. The analysis of PPG for 660nm and 860 nm shows that the mean of power for 860nm was greater than mean of power for 660nm ($p < 0.05$) using t square test in Minitab 17. Oxyhaemoglobin and deoxyhaemoglobin concentration change can be detected by measuring the mean square spectrum for 660nm and 860nm respectively.

General Terms

PPG signal analysis, Paired t Test Statistical Analysis.

Keywords

Cardiac Output (CO), Blood Flow, Stroke Volume, Heart Rate, Photoplethysmography.

1. INTRODUCTION

Heart the vital organ in the cardiovascular system provides nutrients and removes wastes from the body. It pumps the oxygenated blood for the systematic circulation. Cardiac output (CO) for a normal healthy person at rest, is approximately 5 L/min and for an athlete after workout CO exceeds 30 L/min. For circulatory shock patient, CO is than 2 L/min. CO indicates the hemodynamic state of the human. Thus, it would be a tremendous asset to determine CO accurately, reliably, and continuously using minimally invasive methods [1].

The traditional methods of examining the heart using ECG will not give a clear idea about the peripheral blood circulation. Each time the heart beats, a pulse is generated due to the activity of heart which can be extracted from any part of the body by designing an appropriate system. This method discussed in the paper is a non-invasive technique of

measurement and is based on the analysis of the photoplethysmography signal. There are many principles on which the cardiac output measurement techniques are based. They can be categorized into two types: Invasive and non-invasive techniques. Some of the invasive techniques to mention are the thermal convection method, electromagnetic flowmeter, Fick's method and dye dilution techniques [2, 3].

Measurement of cardiac output by:

Flow Meter: CO is the product of stroke volume and the heart rate. Stroke volume is amount of blood ejected by the heart during each beat. Using flow meter we can measure cardiac output directly and accurately. Ultrasound sensor is placed on major vessel and stroke volume is calculated by integrating the area under the pulsatile blood flow. This method is not suitable just for the diagnosis purpose as it requires the invasion and the surgery [3].

Fick Method: In this method for CO measurement the concentration of oxygen in the arterial and venous circulation blood is measured. As the system requires the catheterization the method is invasive.

Frank estimated cardiac output: by measuring total peripheral resistance and mean arterial pressure.

$$CO = \frac{MAP}{TPR} \quad \dots (1)$$

Equation 1 shows the relationship between cardiac output (CO), mean arterial pressure (MAP) and total peripheral resistance (TPR) [3-5].

In thermal Dilution Techniques:

A thermal indicator of known volume is introduced in the right atrium which will produce a temperature change in the pulmonary artery and computing the average cardiac output (CO). Thermal dilution method has a limitation that measurement can be done at certain intervals because of the requirement of administration of bolus stream. The technique requires reliable measurement and well trained operator. Invasive methods cannot be used for all critically ill patients. To overcome this problem of invasive CO measurement techniques like Doppler ultrasound and Transthoracic bioimpedance have been introduced as non invasive CO measurement methods. In ultrasound technique measures the doppler shift in frequency of the beam reflected from the flowing aortic blood. This Doppler technique require expensive equipment and expert operator to stabilized ultrasound sensor. In transthoracic bioimpedance method, impedance change offered by the thorax during each cardiac cycle is measured but this method is less accurate for patients having excessive fluids in the lungs[6].

Estimating cardiac output from arterial blood pressure:

The aortic pressure waveform is governed by the relationship between pressure and flow in the aorta and also by the aortic flow itself. The arterial pressure waveform measured in a peripheral artery arises from the interaction between the arterial system and the heart. In the aorta the viscous effect are negligible so that the blood flow components and pressure waveforms are similar in shape [6].

The problems of invasive measurements are circumvented in an instrument which is based on the analysis of pressure pulse contour, developed by Wessling et al. as shown in fig. 1. They developed an algorithm for stroke volume calculation during systole from aortic impedance and arterial pressure shown in equation 2 [7,8]. As the incident wave reflected from periphery. The wave consist of incident pressure wave which is proportional to stroke volume and reflected pressure wave. The accuracy of cardiac output measurement by pulse contour is affected by total peripheral resistance [9,10].

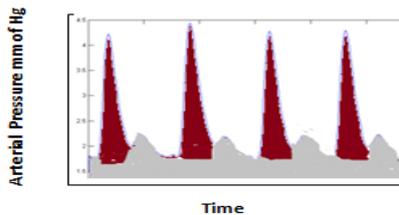


Fig. 1: Arterial Pressure Waveform

$$SV = \frac{\int \frac{dP}{dt}}{Z} \dots (2)$$

Equation 2 shows the calculation of stroke volume using pulse contour analysis. Stroke volume (SV) is estimated from the integral of the change in pressure (P) from end diastole to end systole (t) i.e. the systolic portion of the curve until aortic valve closure. This estimate of stroke volume is also dependent on the impedance of the aorta (Z).

2. PROPOSED METHOD FOR MEASURING THE CARDIAC OUTPUT

The measurement of blood volume changes in the peripheral circulation beneath the skin by means of PPG depends on the fact that blood absorbs light more strongly than skin and tissues. The changes in the blood volume caused by the pressure pulse can be detected by illuminating the skin with a source and then detecting the light reflected by using a detector [5]. The signal obtained from the detector is termed as the PPG signal. This signal is a rich source of diagnostic information which needs to be extracted appropriately.

Stage 1- Sensor Design

The optical sensor developed for the acquisition of the PPG signal is as shown in fig. 2.

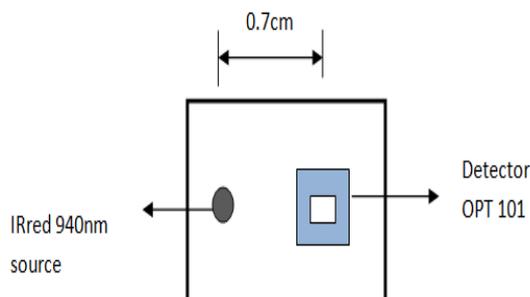


Fig. 2: Optical sensor using 940nm

In present work reflectance type PPG sensor is designed using LED source of 940nm wavelength and detector OPT 101 placed near to each other with a distance of 0.7 cm. To

reduce the effect of ambient light noise signal the polyurethane rubber is used. The current limiting resistors are used to limit the current passing through the LEDs [11]. The light rays from the LED's are reflected and scattered from the skin, tissues and the blood vessels are measured by the detector OPT 101 [12]. Blood flow variation in the vessel shows amplitude variation in the PPG signal.

Stage 2- Acquisition PPG signal

The OPT 101 detector output is low in amplitude it is very difficult to do the acquisition of the PPG signal. The amplification is required as the PPG signal is a low amplitude bio signal and filtering removes the artifacts, disturbances due to noise and movements. The signal amplification is done by using AD620 and noise is removed by low pass and notch filter. Fig.3 shows the signal conditioning circuit.

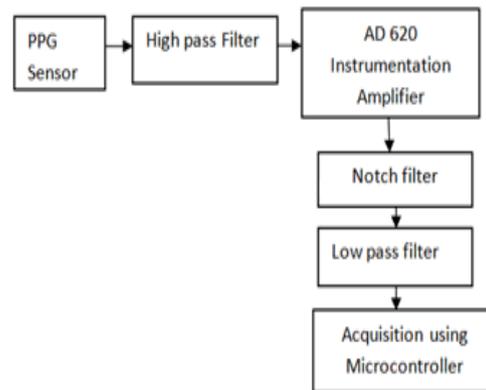


Fig. 3: Blocks of Signal conditioning circuit

The steps involved are:

- The high pass filter designed is used to remove out all the low frequencies from the PPG signal got from the optical sensor.
- Amplification with the instrumentation amplifier AD 620 is the next step of conditioning the signal.
- The signal from instrumentation amplifier is then passed on to the Notch filter for further filtering.
- The low pass filter is next designed for removing out the high frequencies from the PPG signal coming from the Notch filter as shown in fig.4.

Once these steps are applied to the signal it becomes much more suitable for the further processing and data extraction.



Fig. 4: Signal conditioning circuit and the Sensor

2.1.Acquisition of the PPG signal

Acquisition of the signal after amplification is done by using ATmega 8535. ATmega 8535 is a RISC architecture 8-bit microcontroller. 8 channel, 10 bit successive approximation ADC converts the PPG signal into digital signal. The signal after serial transmission is recorded and analyzed in MATLAB. The recorded signal as shown in fig.5 consist of systolic and diastolic peak and dicrotic notch due to closing of aortic valve of heart [8].

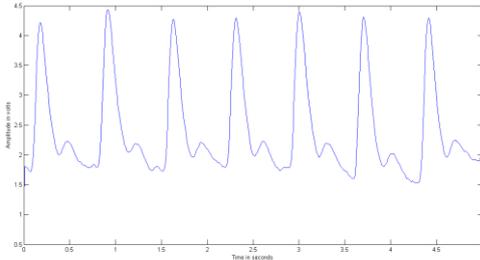


Fig. 5: PPG signal on fingertip

2.2.Analysis of acquired PPG signal on MATLAB

It is this digitized PPG signal that carries useful information which needs to be extracted appropriately. In the work that is carried out, blood flow is the parameter which needs to be derived from the signal. The dependent parameters that are derived from the signal to obtain the blood flow are; stroke volume and heart rate.

The stroke volume is the amount of blood that the heart pumps every beat. This stroke volume can be obtained by estimating the area under the curve of the PPG signal. By averaging out the area for a number of curves obtained for a particular subject the area under the curve can be obtained. The average area so obtained is divided by the patient calibrator Z_{a0} to obtain a quantity approximating the stroke volume. The average value of Z_{a0} is found to be 0.140 for adults [1]. This is depicted by the equation given below:

$$\text{Stroke Volume (SV)} = A / Z_{a0} \text{ (cm}^3\text{)}$$

The instantaneous heart rate is derived as the inverse of the time lapse between the onset of ejection for every two consecutive pulses. This is given by the equation below:

$$\text{Heart Rate} = 60/T \text{ (BPM)}$$

Finally, the blood flow at the site of measurement is computed for each beat as the instantaneous product of stroke volume and heart rate and is given by the equation:

$$\text{Cardiac output} = (\text{SV}).(\text{HR})/1000 = 60.A/1000. Z_{a0} \text{ T (lit/min)}$$

The blood flow so calculated tells about the volumetric changes happening across the site of measurement and is an important parameter for early diagnosis of diseases, monitoring functioning of heart [13-14].

3. EXPERIMENTAL RESULTS

In order to study and draw reliable conclusions from the work carried out experimental analysis was carried out through data collection. Data collection was done on subjects of different age groups and gender. Table 1 summarizes the same.

Table 1. Blood flow recordings of 40 subjects of different age groups

Sl No	Age Group (in Years)			
	15-30	30-45	45-60	60-75
1	4.9	6.68	4.91	4.48
2	5.4	7.9	4.66	3.22
3	6.68	6.89	5.23	4.16
4	7.22	6.39	3.45	4.88
5	5.68	5.22	6.72	5.83
6	6.69	3.62	7.89	5.77
7	6.5	5.2	4.87	5.19
8	5.12	7.48	5.66	5.23
9	7.88	5.55	6.61	4.32
10	8.16	5.31	4.31	5.32
Female				
Male				

From the data collection it is observed that, with the increasing age group the average cardiac output of subjects goes on decreasing. Fig. 6 shows the statistical analysis using MiniTab 17.



Fig. 6: MiniTab Analysis

The readings are compared/verified with the standard Doppler echo method. The examination was performed at Columbia Asia hospital. The differences in the measurement on an average is 0.75 Lit/min.

4. CONCLUSION AND FUTURE SCOPE

From the data collected it is observed that with the increasing age groups the blood volumetric changes happening across the measurement site is decreasing. The calculated correction factor of the developed system is +0.75 lit/min. The average accuracy of the devised system is calculated to be $\pm 2.5\%$. Advantages of the system are non invasive, less expensive method uses optical sensor which is rugged, requires less maintenance thus making the construction simple. Disadvantages of the system: PPG signal is prone to disturbances and noise. Improper or distorted signal results in less accurate results. In the proposed method of measuring cardiac output, an optical sensor is designed to acquire the photoplethysmography signal from the finger non-invasively. Based on the analysis and data extraction carried out on the

signal the cardiac output is estimated. The PPG signal finds numerous applications. It carries volumes of information which needs to be extracted by appropriate methods. It can be used for measuring the cardiac output, blood pressure, vasospastic conditions, heart rate variability and other cardiovascular variability assessments. Understanding the dynamics of the PPG signal can pave the way for many other research works.

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